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(54) Title: MICROCELLULAR POLYMERS AS CELL GROWTH MEDIA AND NOVEL POLYMERS				
(57) Abstract <p>Microcellular polyhipe polymer scaffold suitable for growth of living matter for biomedical applications, obtainable by polymerising a high internal phase emulsion, comprising a homogeneous cross-linked open cellular material defined by a bulk polymer matrix having a surface and an interface with an internal phase, and having porosity greater than 75 % comprising emulsion derived pores of diameter in the range of 0.1 to 10,000 micron and emulsion derive pore interconnects of diameter in the range of up to 100 micron, wherein the scaffold comprises a plurality of discrete and/or interpenetrating zones: at the polymer surface; within its bulk matrix; at the interface between polymer and internal phase; and/or between adjacent but distinct pores and/or interconnects, characterised by form and dimension of pore and interconnect type within each zone, and location of zones wherein adjacent or interpenetrating zones are distinguished by boundaries which may be between or contained within adjacent pores and/or interconnects in respective zones, whereby zones are suitable for regulating positioning and/or morphology of living matter; a biologically active system, or organ support module comprising the scaffold and methods and processes for preparation thereof and use thereof.</p>				

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**MICROCELLULAR POLYMERS AS CELL GROWTH MEDIA AND
NOVEL POLYMERS**

The present invention relates to microcellular polymers as cell growth media,
5 biologically active systems and organ support modules comprising the
polymers and biological cells, a method for cell growth in the materials, the
use thereof as implants in the human or animal body or for *in vitro* studies,
novel processes for the preparation and modification of microcellular
polymers and the polymers obtained thereby and a method for manufacture
10 of an organ support module.

More specifically the present invention relates to microcellular polymers
having porosity in excess of 75%, in the form of a pore and pore interconnect
structure having relative interconnect and pore dimensions d/D as cell growth
15 media for three dimensional cell growth throughout the polymer, the system
comprising this, method for growth and use thereof, and novel processes for
the preparation thereof and polymers obtained thereby.

Microcellular synthetic organic/inorganic/natural materials in which the
20 pores are interconnected are often used as a support to grow plant or animal
cells or enzymes.

Tailoring of a microcellular support for a specific cell growth application is
often difficult with naturally forming microcellular materials. Recently
25 efforts have been made to grow cells on newly developed microcellular
materials, which are prepared through a high internal phase emulsion (HIPE)
polymerisation route in which the phase volume of the dispersed phase is
greater than about 75% and subsequently polymerising and/or cross linking

to obtain a rigid structure, offering many advantages. These materials have been described both in open literature and in patent disclosures, and they are referred to as polyhipe polymers (PHPs).

5 It is stated that the important features of these materials are:

1. The pore volume can be as high as 97% (for practical purposes);
2. Pores are interconnected;
3. Pore and interconnect sizes can be controlled accurately and
10 independently;
4. Additional polycondensation reactions can be achieved;
5. Crosslinking can be achieved with proteins, polymers, silicates or organic polymers plus a wide range of porous materials can be obtained;
- 15 6. Post polymerisation/crosslinking modification of PHP is possible which is further facilitated by using functional monomers at the emulsification stage;
7. PHP can be obtained in block and/or in particulate form;
8. Since HIPEs can be pumped, it is possible to form moulded structures.

20

US patent 5,071,747 describes the preparation of microcellular polymeric support materials with average void (pore) diameter within a range of from 1 to 150 μm interconnected by holes (interconnects). Hole size (d) is related to the void size (D) and their ratio can be controlled in the range of $0 < d/D <$
25 0.3. The control of d/D ratio is through the control of the surfactant concentration and by the addition of electrolyte, mainly CaCl_2 in the range of 10^{-4} molar to 5 molar. The electrolytes are selected from soluble halides

and sulphates. The function of the electrolyte is claimed to be the control of the size of the holes and to improve stability of the emulsion.

In stage-1, the oil and aqueous phases are introduced under deformation (agitation) and in stage-2, the resulting HIPE is homogenised under deformation (agitation). Although the rate of introduction of the phases (i.e. dosing time (t_D)) of the dispersed phase into the batch mixer, and the subsequent homogenisation time (t_H) of HIPE is quoted in US patent 5,071,747, the mixing conditions are not specified. Polymerisation is followed by introduction of cells.

US 5,071,747 discloses isotropic (non-directional) plant and animal cell growth in three dimensions (3D) in polyvinyl Polyhipe Polymer (PHP), the generic name for the micro porous material. Reactants or nutrients are provided via inter connecting holes (interconnects) within the PHP to access voids (pores) in which cells are grown or reacted to produce products. The PHP is used purely as a cell growth medium allowing normal cell functioning within voids, using holes (interconnects) as access channels. Voids and holes were provided having void diameter of 6-12 times the diameter of cells to be introduced and grown and hole diameter of 3-6 times, i.e. void diameter 45 micron and hole diameter 15 micron in the case of growth of yeast cells of the order 5 micron. In these cases, however cell growth was predominantly surface growth. In a further example PHP having porosity of 90%, void and hole diameter of 30 micron and 10 micron respectively was used for fungal growth which was found to penetrate the porous polymer network.

Nevertheless the literature gives only limited teaching with regard to controlling pore and interconnect size. In US 5,071,747 above it is not stated how the claimed dimensions were achieved and in fact it would not be possible to obtain pore dimensions in excess of 50 microns using the limited information given, nor would it be possible to control or pre-determine a particular pore size within the range 1-50 microns without extensive experimentation.

The usefulness of PHP materials for cell growth remains therefore extremely limited and there is a need for materials and methods for preparing more sophisticated polymers enabling more sophisticated cell growth.

We have now surprisingly found that coherent cell growth may be achieved within PHP polymers to provide a co-operating multi cell system which is adapted for a number of uses and that novel PHPs may be obtained having novel properties adapted for biological and non-biological uses. It is particularly advantageous that cells co-operate in the PHP of the invention as this is important for cell growth.

Accordingly in the broadest aspect of the invention there is provided micro cellular polyhipe natural or synthetic polymer in the form of a homogeneous cross linked open cellular material having porosity greater than 75%, comprising pores of diameter 1 to 10,000 micron and pore interconnects of diameter of up to 100 micron, wherein the polyhipe is a scaffold for multicell growth in three dimensions wherein the polymer pores and interconnects are comprised in a plurality of distinct or interpenetrating zones which are adapted to regulate cell positioning and morphology whereby cell growth is constrained within and/or extends throughout plural zones in directional

and/or nondirectional manner to provide a multiple cell structure for biomedical applications.

The polymer according to the invention may be provided with additional
5 features in the form of micro capillary networks, nanopores, surface, chemical modification, electrical or like properties for specific purposes.

The invention thus provides useful PHP materials, the many applications of which are highly significant specifically the invention can be used as a multi-
10 channel in-vitro organ module in which the cells must be under a certain chemical and/or electrical potential, to express and thus function. The chemical and/or electrical potential will be provided by the contents of the micro-channels. This type of module is necessary in organs like liver, kidney and pancreas. Modules with micro-channels can also be used as selective
15 bio-reactors in which the cell expression is regulated through the presence of a chemical and/or electrical potential imposed through each type of channel content. Cell expression under different types of potential and potential strength can be used to produce biochemicals which can lead to the understanding of disease processes and/or preparation of new proteins in
20 drug applications.

The use of PHP materials in the production of biochemicals is not confined to the animal cells grown on a support system with a preferred architecture and physiological conditions. Plant cells as well as micro-organisms such as
25 bacteria and virus can also be grown using the PHP material. In this case particulate form of the PHP material with open pores can be used to provide support for cell, bacteria or virus growth. The particulate PHP material can be suspended in a suitable biomedica for the biochemical reaction to proceed.

Monolithic support systems such as those described in this invention can also be used for this purpose also. When viruses are to be grown on PHP support, small pore size material, less than 1 micron, may be preferred.

- 5 Preferably zones are adapted for growth of multiple cell types independently constrained within and/or extending throughout multiple zones respectively.

The important parameters that promote cell growth are the pore and interconnect sizes and chemical and physical characteristics of the support surface. In some cases, the cell growth may be anisotropic (directional) and therefore the pores of the cell support may be required to be in the form of micro-channels with interconnects to provide cell-communication and cell penetration. The walls of the micro-channels can be (bio)degradable so that subsequent cell fusion can be obtained after (bio)degradation.

15

Biomedical applications include any application in which the materials interface with biological systems to evaluate, treat, augment or replace any tissue, organ or function of the body. The materials may therefore be envisaged for a range of applications for example the manufacture of contact lenses, dental fillings, cochlea implants, vascular supports including heart valves and cardiac pace makers and drug delivery skin patches and the like.

Reference herein to a scaffold is to a porous material which provides support and allows a positioning or load bearing function on the scaffold at an early stage as the scaffold directs growth or takes the load until cells are grown within a scaffold and have developed extracellular matrix to gain load bearing ability and with stand loads themselves.

25

Cell growth may be of cells which are introduced into the scaffold or migrate into the scaffold, for example in the case that a scaffold is used for synthetic growth or is used for natural cell growth by migration of cells from surrounding tissue.

5

Reference herein to polyhipe is to any natural or synthetic polymer comprising pores and interconnects as hereinbefore defined obtained by polymerising a high internal phase emulsion.

- 10 Reference herein to pores and interconnects in the polymer is to empty pores or cells with pore interconnects therebetween, which may be empty or may contain dissolved or dispersed materials. Pores and interconnects are distinguished by their relative magnitude and dimensions as herein below defined. The total porosity of the material is the combined empty space
15 provided by the sum of all pores and interconnects.

Reference herein to zones within the polymer is to distinct or interpenetrating regions characterised by the form, location, magnitude or other property of pores and interconnects comprised in the zone. For example one or more
20 zones are provided at the polymer surface, within its bulk matrix, at the interface between polymer and internal phase, between adjacent pores and/or interconnects of different form or dimension or adapted for growth of different cell types. Zones are distinguished by boundaries which may be between or contained within adjacent pores and/or interconnects in
25 respective zones. Cell support material with two or more distinct zones where the pore and interconnect sites are different will be needed when co-culturing two or more cell types.

Preferably polymer according to the invention is suited for growth of multiple cell types to provide a multi zone cell structure in which selected cell types are confined to the specific boundary whilst other cell types grow throughout the structure across boundaries presented by the scaffold. By this means the polyhipe scaffold provides for a multiple cell system extending throughout the scaffold, optionally including intrainterconnect or intramicrocapillary growth and exhibiting zoning and optionally interzoning of biological cells.

The polymer of the invention is moreover able to encourage natural zoning of cells by migration within, throughout and between zones which have been tailored for preferential migration of desired cell types.

The polymer of the present invention may be obtained from any desired natural or synthetic monomers, oligomers, macromonomers, reactive polymers and mixtures thereof which exhibit bio-compatibility. Polyhipes are commercially available or may be prepared using methods as disclosed in US 5071747 and in additional patent publications referred therein or as hereinbelow described.

20

The generic polyhipe polymer which is commercially available comprises polyvinyl polyhipe and is made up of oil phase monomers styrene, divinyl benzene (DVB) and surfactant (Span 80 sorbitan monooleate), and may be in rigid or flexible form depending on the relative proportions of monomers, additionally in flexible form including monomer 2-ethylhexyl acrylate, and in the aqueous phase an amount of potassium persulphate as aqueous phase initiator.

The polymer of the invention may be natural or synthetic, soluble or insoluble, optionally (bio)degradable crosslinked polymer, preferably selected from proteins and cellulose, polyacrylamide, polyvinyl in rigid or flexible form, poly(lactic acid), poly(glycolic acid), polycaprolactone, poly
5 (lactide/glycolide) and polyacrylimide.

Commercially available polyhipe typically has pore diameter in the range 5-50 micron.

10 PHP support according to the invention may have pore diameter in the range 0.5 or 1-10,000 micron. A preferred lower range limit is 10 micron, more preferably 30, for example 50 micron. A preferred upper range is 300 micron, more preferably 200 micron. Polyhipe of particular pore diameter may be obtained by methods described hereinbelow, and may have any
15 desired ratio of interconnect to pore diameter, for example in the range $0 < d/D < 0.5$, preferably in the range $0.1 < d/D < 0.5$ when the pore diameter is approximately less than 200 micron. Interconnects may have diameter in a range of up to 100 micron, preferably 0.001 to 100 micron, more preferably 10-50 micron. Extensive networks of elongate micro capillaries of diameter
20 as low as 10 micron may be provided, these capillaries being separated by the microcellular polymer. A zone comprising the interface between a capillary wall and the bulk polymer may provide a surface layer of any desired thickness for growth of cells of any desired size. An interface having a thin surface layer of the order of 0.5-5 micron is particularly suited for
25 growth of neurons (nerve cells) or muscle cells in which directionality is important. The interface having smaller pore size than the bulk provides an ideal zone for growth of cells forming a lining, for example cells lining the blood vessels or for growing endothelial cells on the interface surface.

Pore structure in the present invention are formed through 4 different mechanisms. These mechanisms are used, often in combination, to obtain several forms of support architecture in order to create a realistic model for organ support. Four different pore structures are described below:

Type - 1 Pores (Basic pores): This is the basic pore structure the size of which is determined at the emulsification stage of the PHP formation. Therefore, the pore size is mainly determined by the deformation (flow) history of the emulsion. The integrity of these pores are kept during polymerisation and the interconnects are formed at this stage. Depending on the chemistry of the oil and aqueous phases, phase volume and the polymerisation conditions such as temperature and pressure, the interconnect size can be controlled in the range $0 < d/D < 0.5$.

15

Type - 2 Pores (Coalescence pores): This type of pore architecture is obtained through the controlled coalescence of the Type-1 pores during polymerisation. The dispersed phase droplets in the emulsion of the PHP are coalesced by the addition of water soluble polymers into the aqueous phase, or by adding slightly hydrophilic oils (such as styrene oxide) to the oil phase. The interconnect size in this case is the same as that of the Type-1 pores which form a matrix incorporating the coalesced pores. However, due to the fact that the coalesced pores are very large compared with the basic pores, the d/D ratio is very small.

20

Type 3 - Pores (Micro-capillaries): It is possible to obtain a network of microcapillaries within the Type-1 or Type-2 pores (i.e. supported by these pores) or indeed to obtain microcapillaries supported by Type-1 pores with

Type-2 pores forming the walls of the capillaries or vice versa. These microcapillaries are formed using a specially constructed mould, in which rods or fine fibres are inserted. The PHP emulsion is pumped into the mould and polymerisation is carried out. After polymerisation, these inserts are removed either by pulling them out (when polymerisation results in shrinkage, or when the polymeric inserts are first swollen during polymerisation and subsequently shrunk upon polymerisation or the removal of the solvent used as filler for the monomer) or by dissolving the inserts in acid or a suitable solvent. The minimum diameter of these micro capillaries can be as low as 10 micron (using glass, carbon or polymeric fibres). Metallic fibres can be used to obtain microcapillaries in the range 50-1000 micron or greater.

Type - 4 Pores (Nano-pores): The pore walls of the microporous PHP can be made nano-porous by using a suitable 'filler' in the oil -phase so that, after polymerisation, the filler or its reacted form can be removed by solvent extraction. The filler can be oil or another monomer or macromonomer which do not undergo polymerisation or if the monomer/macromonomer polymerizes, the resultant polymer could be extracted. The filler can also be another polymer or reactive polymer which dissolve in the monomer or a suitable hydrocarbon oil as well as nano-sized solids such as silica powder or zeolite. If the 'filler' oil is highly soluble in the monomer as well as the resulting cross linked polymer, it will create very fine pores after extraction. If the resulting polymer is not soluble in the filler oil or if it does not swell, they will form sub-micron particles attached to each other by a number of crosslinking chains. This network of particles and the filler oil will form a co-continuous system. This method can be used to control the amount of nano-porosity in the micro-porous polymer. Nano-pores are also

obtained by using polymers which are soluble in the monomer. These polymers can also be extracted after the formation of the polyhipe polymer to create nano-pores. Such filler polymers, if not extracted, can provide extra strength. Such nano-pores will allow the diffusion of small molecular weight solutes across the organ support since the walls will create a barrier for the transport of these vital solutes. Such solutes include oxygen, carbon dioxide, nutrients, growth hormones, electrolyte and the like. The presence of the nano-pores can also be useful in the controlled biodegradation of the organ support through hydrolytic polymer degradation since the effective surface area will be very high.

Pore types may be present in any modular form, for example shell and tube type or cubic/polyhedral.

Preferably pore size and inter connect size are selected according to the type of cell to be grown and the type of growth, i.e. with or without penetration, confined to a boundary or crossing boundaries between zones. Pore size in zones intended for growth of cells throughout the zone is preferably of diameter 2 – 3 times the diameter of cells to be grown, for example 2.5 times cell diameter. Without being limited to this theory, it is thought that this ratio of dimensions is optimum for cell types, including cartilage cell types, whereby cells are able to grow and prosper, as indicated by production of collagen. For example cartilage cells of diameter 10 micron were grown in pores of diameter in the range 17-30 micron, specifically of 25 micron.

The polyhipe scaffold may be adapted to provide a desired surface characterised by means of the surface coating, using coating materials introduced *in situ* during polymerisation or post polymerisation. Any known

materials typically used for coating polymers for use in cell growth may be employed, for example cell growth promoters such as hydroxylapatite or tricalcium phosphate, which also promotes biocompatibility, other minerals, silica, collagen, hyaluran, polyethylene oxide, carboxymethyl cellulose (CMC), proteins, organic polymers, particles and the like. Coating *in situ* provides homogeneous coating throughout the scaffold. Post polymerisation coating, for example using laser ablation results in coating confined to the surface of the bulk matrix and may be particle coating.

The polyhipe scaffold may be constructed of any desired materials as hereinbefore defined. It has been found that cyclic mechanical straining of the support or the application of an electric field may also enhance the rate of cell growth. In most cases, the biodegradability of the support does not appear to be important and therefore the support material with growing cells can be implanted in the animal body without fear of rejection. However, when cell-cell fusion is required, (bio)degradability becomes important.

Therefore the polymer is preferably made of resiliently deformable or elastic materials or is rendered resiliently deformable or elastic by suitable means. Preferred materials are therefore thermoplastics which may be deformed in suitable manner to influence cell growth. Preferably a polyhipe support as hereinbefore defined is suited for repeated stress and relaxation by means of oscillatory straining of the scaffold during cell growth. This has been found to have beneficial effects in cell growth rate promotion.

The polyhipe scaffold as hereinbefore defined may be electrically conductive or may be rendered electrically conductive by known means whereby it is adapted to conduct an electric current during cell growth. This technique is

particularly advantageous for distinguishing certain cell types and promoting growth and fusion of particular cell types such as neurons and muscle cells.

- The polyhipe scaffold may be biodegradable or may be rendered
5 biodegradable by known means whereby it is adapted to be degraded by contact with any desired agent or by introduction into any desired environment, for example a specific biological agent or environment. Preferably the polyhipe scaffold is adapted for degradation by the hydrolysis or enzyme-catalysed hydrolysis of the polymer chains and cross link or by
10 the erosion of the polymer. By this means the polyhipe scaffolds may be used as a scaffold to support a biologically active system for the duration of time required for that system to establish itself and be self supporting and thereafter degrade with no detrimental effects. Preferably a degradable scaffold is used for growth of cell types including for example muscle cells.
15 By this means a scaffold is adapted to position multiple cell types in pre determined zones and thereafter to degrade allowing self fusion to take place in the absence of the intervening scaffold whereby functioning biologically active micro systems are created.
- 20 In a further aspect of the invention there is provided a polyhipe scaffold as hereinbefore defined comprising multiple cells characterised by growth in three dimensions as hereinbefore defined in polymer zones as hereinbefore defined.
- 25 The scaffold including cells may comprise any of the properties as hereinbefore defined.

Multiple cells may be any desired cell type selected from human, animal and plant cells. Preferably cells are human or animal cells and are representative of multiple cell types present in any organ, system or part of the human or animal body. Cells are preferably selected from isotropic tissue and bone
5 cells present in cartilage, cornea, marrow and the like, anisotropic cells such as nerve, muscle, blood vessel cells and the like. Cell type includes for example fibroblasts, chondrocytes, osteoblasts, bone marrow cells, hepatocytes, cardiomyocytes, neurons, myoblasts, macrophages and microvascular endothelium cells.

10

In a further aspect of the invention there is provided a biologically active system comprising a polyhipe scaffold and multiple cells as hereinbefore defined adapted to provide normal cell functioning associated with a natural biologically active system present in the human or animal body. A
15 biologically active system may comprise the polyhipe scaffold intact or in partially degraded state as hereinbefore defined adapted for fusion or assimilation into an environment.

In a further aspect of the invention there is provided a method for growth of
20 multiple cells in a polyhipe scaffold as hereinbefore defined comprising providing cells on or in the scaffold in a controlled environment and providing a suitable nutrient adapted for growth in known manner. The method of the invention may include any technique for growth promotion, positional control and the like adapted to provide a structural system as
25 hereinbefore defined.

In a further aspect of the invention there is provided a polyhipe scaffold adapted for multiple cell growth as hereinbefore defined, comprising

multiple cells as hereinbefore defined or a biologically active system as hereinbefore defined for use as an implant *in vivo* in the human or animal body or as modules for *in vitro* studies mimicking a part of the human or animal body or for use in a growth environment for further systems as
5 hereinbefore defined, for example for the growth of organ cells in the cell side of the module in order to simulate organs.

In a further aspect of the invention there is provided an organ support module comprising a cubic or polyhedric module of closely interwoven but not
10 interconnecting channels immersed in PHP. The module is suited for growth of specific organ cells in the PHP and/or the channels. Some cells may be in contact with a specific microchannel and all cells will be capable of intercell communication.

15 According to this aspect of the invention *in vitro* organs may be produced using the micro capillary interconnects of the scaffold as hereinbefore defined for nutrient circulation and waste product removal while growing cells in the bulk pores and additionally or subsequently when anisotropic cell growth is needed for example to mimic cardiovascular systems and the like,
20 employing circulation of nutrients and waste products removable through the bulk pores of the micro porous scaffold while growing cells in the micro capillary interconnects. It is a particular advantage of the invention that this allows growth and sustains cell functioning for prolonged periods. The use of the scaffolds and systems as hereinbefore defined as *in vitro* organs may
25 be harnessed for testing of pharmaceuticals and any substances to be introduced into the human or animal body, for genetic engineering, for studying mechanisms associated with disease and the like under controlled conditions and the like. This is particularly advantageous in eliminating the

need for animal testing. *In vitro* modules as hereinbefore defined for specific types of multiple cells and applications may be provided and may be associated with instrumentation to provide the necessary physiological conditions.

5

The module is suited for growth of specific organ cells in the PHP and/or the channels. Some cells may be in contact with a specific microchannel and all cells will be capable of intercell communication.

10 In order to mimic the cells natural environment within an organ, a number of facilities have to be provided for the physiologically correct cell function. These facilities can be summarised as:

1. blood or nutrient supply microcapillaries;
- 15 2. cell expression collection channels;
3. neural stimulation;
4. microcellular support for the co-culture of support cells;
5. biocompatibility and controlled biodegradability; and
6. controlled mechanical, chemical and electrical characteristics of
- 20 support.

It is possible to allocate a set of micro-capillary channels which are separated by cell support material (bulk of the organ support) to facilitate (1)-(3). In such an organ support system, cells are within a few hundred micron of these

25 micro-capillaries. In each set of capillary network as well as in the bulk, mean pressure can be regulated so that the cell expression products are collected in the designated set of capillaries and subsequently recovered. Neurons can be grown in the designated set of micro-capillaries and

electrically stimulated through an external micro-chip. Since the cell expression is affected by the external stresses (physical stresses such as cyclic strain and temperature, chemical stresses such as drug/toxin in blood/nutrient, electrical stimulation through the nerves) it is possible to
5 obtain different cell expressions depending on the external stress which can be controlled very accurately in an *in vitro* organ. *In vivo*, the range and duration of external stresses are limited and there is always interference from the other cells. Therefore, *in vitro* organs with physiological facilities are useful to extend the scope of cell response to external stresses. In this
10 respect, *in vitro* organs can be viewed as Biomedical Reactors which allows the synthesis of novel proteins and understanding of disease processes.

One set of capillaries can be used for seeding so that cells do not need to travel a long distance in order to occupy the available space. After seeding,
15 these micro-capillaries can be used for other purposes.

It is possible to design an organ support system which allows the co-culture of support cells in close proximity to the main organ cells. PHP emulsions with differing pore and interconnect sizes can be co-extruded in a desired
20 configuration (i.e. concentrically or side-by-side). As found in this invention, cells do not penetrate if the pore and interconnect sizes are too large or too small with respect to the cell size. Therefore, each type of cell will be able to choose the optimum pore structure for support and growth.

25 The electrical stimulation of growing nerve cells in a micro-capillary can be achieved by coating the micro-capillaries with conductive polymers. In the absence of any neural network for electrical stimulation, controlled electrical

fields within the support system can be created by using a network of carbon fibres as conductors.

It should be appreciated that the invention derives from the finding that polyhipe scaffolds as hereinbefore defined may be provided with controlled pore dimensions and diameters in zones as hereinbefore defined, such that for example one cell type penetrates the scaffold and a second cell type may grow on the surface. This enables controlled cell growth and positioning in manner to create a multiple cell structure which is adapted to co-operate to provide biological activity.

In a further aspect of the invention there is provided a process for the preparation of microcellular polyhipe natural or synthetic polymers as hereinbefore defined comprising in a first stage the formation of a high internal phase emulsion (HIPE) of dispersed phase in continuous phase, wherein the dispersed phase may be void or may contain dissolved or dispersed materials, and monomers, oligomers and/or pre-polymers are present in the continuous phase, homogenisation and polymerisation thereof, by means of in the first stage introducing the dispersed phase by controlled dosing into the continuous phase with controlled mixing at controlled temperature to achieve an emulsion, and subsequently homogenising for controlled period under controlled deformation and polymerising, under controlled temperature and pressure.

The process of the invention enables a vast range of pore sizes to be obtained by controlling processing technique and compositional conditions and in particular as hereinbefore defined to obtain Type 1-4 micro-pores. The pore

sizes are obtained in roughly 3 groups: they are; small pore size: 1-10 μm ; large pore size: 11-200 μm and very large pore size: 201-10,000 μm .

Very small pore size emulsions (approaching 0.5 μm) are obtained using
5 very high deformation rate flows in which the flow is predominantly
extensional and the emulsification temperature is as low as possible. Large
pore size emulsions (approaching 200 μm) are obtained at high temperatures
and just above the critical deformation rate below which the emulsion will
fully or partially invert, for example to an oil-in water type system. The
10 critical deformation rate may be determined by varying, for example the rate
of addition or rate of deformation during mixing for a given system. These
emulsions should also be processed in a short time using predominantly
shear flows.

15 Very large pore emulsions (approaching 10,000 μm) are obtained through the
method of controlled pore coalescence during polymerisation. There are two
methods of achieving controlled coalescence: 1) by adding into the aqueous
(dispersed) phase a known amount of water soluble polymer or 2) by adding
'filler' solutes into the continuous oil phase. In both methods, the
20 concentration and the type of these additives are important. If the
concentrations are low, these additives result in polyhipe polymers in the
range of 1-200 μm with some desired properties. If the concentration is
above a certain value, the coalescence pores start to form. In this case, the
pore size is dictated by the size of the pores before the start of coalescence,
25 temperature of polymerisation and concentration, molecular weight and type
of additive.

The emulsion may be obtained from any desired immiscible phases forming a continuous and a dispersed phase, preferably from aqueous and non-aqueous phases, more preferably aqueous and oil phases. The emulsion obtained may be an aqueous in oil emulsion or oil in aqueous emulsion.

5

The process according to the invention may be used for the preparation of any desired polymers as hereinbefore defined.

According to this aspect of the present invention, it has surprisingly been
10 found that by means of controlled dosing the dispersed phase into the continuous phase, it is possible to achieve the desired emulsion. In a batch mixer, dosing of the dispersed phase is preferentially conducted from the bottom of the mixer, using either single or multiple entry points. Multiple entry feed resulted in larger pore emulsions. If the dosing rate was very fast,
15 mixing created by the emerging jet of aqueous phase was too severe and therefore the emulsion pore size decreases. Therefore, this combination of multiple-feed points with a relatively prolonged dosing created large pore emulsion. After the completion of dosing, the emulsion should be homogenised but if the homogenisation period was long, pore size decreased.

20

Controlled mixing as hereinbefore defined may be critical or extended. Critical mixing is sufficient mixing to cause the dispersion of aqueous phase into the oil phase without phase inversion. Critical mixing is obtained by use of a homogeneous mixing field whereby pore size is substantially uniform
25 preventing emulsion breakdown and phase inversion.

Mixing may be by any means suited to provide a homogenous mixing field substantially throughout the total volume of the two phases and is preferably by multiple blade, multiple jet and the like mixing.

- 5 It has surprisingly been found therefore that, contrary to the teaching of US 5,071,747, the achievement of a stable emulsion with large pore diameter is obtained by minimising the intensity of mixing. According to the invention it has been found that by means of dosing, homogeneous mixing and the like a stable emulsion may be obtained dispensing with the need for intense
10 mixing.

Dosing, emulsifying and homogenisation may be conducted at any suitable temperature depending on the pore size in the final polyhipe polymer. If the desired pore size is large, the preferred temperature is high, but nevertheless,
15 below the boiling point of the lowest boiling phase. An aqueous phase as the lowest boiling phase, boils at around 100°C. It has been found that the process of the invention employing emulsification temperature of 60°C or greater results in polymers being obtained having pore size in excess of 60 micron. Increase of emulsification temperature above 60°C results in
20 dramatic increase in pore size by an amount greater than that achieved for a similar increase in temperature below 60°C.

The maximum emulsification temperature may be greater than the normal boiling temperature of the lowest boiling phase, for example the lowest
25 boiling phase may include any suitable component adapted to raise the boiling point. Preferably the aqueous phase includes an electrolyte which is stable at 100°C and is potentially inert. By this means the process is

preferably carried out with use of homogenisation temperature in the range 60-150°C, more preferably 80-140°C, most preferably 80-120°C.

Emulsification and subsequent polymerisation can be carried out at
5 temperatures above the normal boiling point of the aqueous or continuous phase materials, by increasing the pressure above atmospheric using closed continuous processing equipment.

The process may be carried out with use of additional aqueous or oil phase
10 initiators, cross linking agents, fillers and the like and it is preferred that these are stable at the maximum operating temperature as hereinbefore defined. Selection of initiators, cross linking agents and the like is made with reference to the acceptable viscosity of the phases for emulsifying and homogenising. It may be acceptable to reduce the amount of cross linking
15 agent required by use of a proportion of pre polymers and partly cross linked pre polymers, optionally with the use of a suitable oil phase filler to increase oil phase volume and reduce effective viscosity.

Preferably the process of the invention is characterised by use of an initiator
20 in the oil phase, together with or in place of an aqueous phase initiator as known in the art.

Preferably the process is carried out with use of oil phase fillers and allowing operation at high emulsification temperature and with use of minimum cross
25 linking agent. This has the further advantage that oil phase filler such as high boiling point hydrocarbon may be leached out after polymerisation creating nano-pores and increasing the size of the interconnects.

Electrolytes may be for example calcium chloride or minerals such as hydroxyapatite.

Aqueous phase initiators may be employed for operation at lower
5 temperatures and include sodium or potassium persulphate.

For operation at elevated temperature above 80°C oil phase initiators are preferably used for example, 1,1-azobis (cyclohexanecarbonitrile).

10 Cross linking agent may be for example divinylbenzene (DVB). If the polyhipe is required to be biodegradable, hydrolyzable crosslinks can be obtained. These crosslinking agents are ethylene diacrylate, N-N'-diallyl tartardiamide, N-N (1,2, dihydroxyethane) - bis -acrylamide, and N-N'-N'' - triallyl citrictriamide. However, in this case of biodegradable crosslinkers,
15 the polymer itself should be biodegradable. These polymers are poly (lactic acid), poly (glycolic acid), poly ε-caprolactam and polyacrylimide.

When water soluble polymers are needed to form the microcellular structure, they need to be crosslinked . In this case of such polymers, monomer (such
20 as acrylamide) is dissolved in water and a HIPE emulsion is formed dosing this monomer solution into a hydrocarbon liquid such as hexane or toluene in the presence of suitable surfactant, initiator and crosslinker.

Proteins and cellulose can also form microcellular structures. In this case,
25 these materials together with a suitable emulsifier are dissolved in a suitable aqueous phase (water for proteins and Schweitzer's reagent, $\text{Cu}(\text{NH}_3)_4(\text{OH})_2$ for cellulose) and dosed into a hydrocarbon liquid to form a water continuous

HIPE. Crosslinking is achieved by immersing the HIPE into a solution of glutaraldehyde (for proteins) or acid solution (cellulose).

Oil phase filler may be any high boiling point hydrocarbon, another
5 monomer, or macromonomer, reactive or inert polymer, solid particles, or
their combinations.

The process may include introduction of any suitable modifier as
hereinbefore defined prior to or subsequent to polymerisation. For example
10 minerals such as hydroxyapatite may be introduced in the dispersed aqueous
phase and dosed into the continuous phase as hereinbefore defined.
Alternatively a post polymerisation modification stage is employed, which
may simply take the form of removing surfactant, electrolyte and unreacted
monomer, coating, further polymerisation or reaction on the existing polymer
15 surface. Modifying agents and modification techniques are shown in the art.

Polymerisation is carried out under known conditions of time and
temperature for the respective monomer, oligomer and or prepolymer to be
polymerised, as known in the art.

20

In a further aspect of the invention there is provided a microcellular polyhipe
natural or synthetic polymer in the form of a homogeneous cross linked open
cellular material having porosity in excess of 75% comprising pores and pore
interconnects formed by polymerising a high internal phase emulsion (HIPE)
25 as herein before defined wherein average pore diameter is in excess of 50
micron. Preferably average pore diameter is in excess of 100 micron, more
preferably in excess of 150 micro and is selected to be suitable for the

desired polymer application. Pore diameters up to 10,000 micron are envisaged. Interconnect diameters are as hereinbefore defined.

5 In a further aspect of the invention there is provided polyhipe polymer as hereinbefore defined having average pore size as hereinbefore defined in the range 1-10,000 micron, wherein the polymer is modified during polymerisation thereof or post polymerisation to be electrically conducting, degradable, or comprises mineral distributed throughout the matrix or as a surface coating.

10

In a further aspect of the invention there is provided an apparatus for the preparation of a polymer according to the invention comprising a mixing vessel adapted to contain the continuous phase having multiple inlets for the introduction by dosing of the dispersed phase as hereinbefore defined, and
15 comprising means for homogeneous mixing as hereinbefore defined and comprising temperature elevating and regulating means.

In a further aspect of the invention there is provided a method for manufacturing an organ support module as hereinbefore defined. The
20 method employs one or more stages as hereinbefore defined for creating Type 1-4 pores.

The invention is now illustrated in non limiting manner with reference to the following examples and figures wherein:

25

Figures 1a – 2b are schematics and a cross-section of organ support systems employing the invention.

Example A - preparation of polyhipe, using apparatus according to the invention.

The preparation of the emulsion was carried out in a batch mixer from an oil phase and an aqueous phase, dosed at a predetermined rate while the emulsion was stirred at constant rotational speed. Dosing rate, deformation rate, mixing rate were predetermined as a function of volume of respective phases, diameter of the batch mixer and of the impellers, rotational speed of the impellers and homogenisation time.

10

In order to eliminate the differences in performance of different mixing conditions we characterise the mixing through:

Dosing rate $R_D = \frac{V_A}{t_D}$

15

Deformation rate during dosing $R_E = V_A / (t_D V_0)$

Mixing rate $R_M = D_i \Omega / D_0$

20 Where:

V_A = Volume of aqueous phase added over a period of time t_D

V_0 = Volume of the oil phase placed in the batch mixer

D_i = Diameter of the impellers

D_0 = Diameter of the batch mixer

25 Ω = Rotational speed

We also define t_H as the homogenisation time and t_T as the total mixing time.

$$t_T = t_D + t_H$$

After the preparation of the emulsions under a given set of conditions as described in each of the following examples, the emulsion was allowed to polymerise at 60°C for 8 hours. Samples were further modified as described in the following examples.

5

The pore and interconnect sizes were determined using scanning electron microscope, using thoroughly dried and washed samples.

10 **Example A1 - the effect of operating condition on the pore and interconnect size.**

The oil phase contained 78% styrene, 8% DVB monomer and cross-linking agent and 14% Span 80 surfactant sorbitant monooleate while the aqueous phase contained 1% potassium persulphate. Two flat paddle impellers (8 cm in diameter and 1.4 cm in width) were used in a mixing tank of 8.5 cm diameter. Impeller separation was 1 cm. 25 ml of oil phase is placed at the bottom of the tank and 225 ml of aqueous phase was dosed using 16 feed points. Temperature of the aqueous phase was ranged from -1.0 to 80°C.

20

25

Table 1 - The effect of operating conditions on the pore and interconnect size.

Temp (degC)	Pore Size(D) (μm)	Interconnect Size(d) (μm)	Dosing time(s)	Homogenisation time(s)	d/D
5	34	9	40	60	0.26
25	37	14	40	60	0.38
60	65	22	40	60	0.33
80	141	37	40	60	0.26
25	102	25	40	20	0.25
25	72	21	30	20	0.29
25	67	22	25	20	0.33
25	148	43	60	20	0.29

5 **Example A2 – Effect of water soluble polymers on pore and interconnect size in controlled pore coalescence.**

Emulsions were prepared using 78% styrene, 8% DVB and 14% Span 80 as the oil phase. The aqueous phase contained 1% potassium persulphate and
 10 varying amounts of water soluble polymers, sodium carboxy methyl cellulose (CMC) or polyethylene oxide (PEO). Emulsification is at 25°C, water phase volume is 85% and the aqueous phase is delivered through a single inlet. The processing conditions were: $t_D = t_H = 600$ seconds, $R_D = 0.70$ ml/s, $R_E = 0.0067$ s⁻¹ and $R_M = 4.7$ s⁻¹.

15

20

Table 2 - The effect of water soluble polymers on pore size in controlled pore coalescence

Relative Molecular Mass	Water Soluble Polymer Concentration wt% of aqueous phase	Pore Size (μm)
Control	0	18
Sodium carboxymethyl cellulose (CMC)		
90,000	0.5	22
90,000	1.0	260
250,000	1.0	1200
Polyethylene oxide (PEO)		
200,000	1.0	420
400,000	1.0	4300

Example A 3 - In-situ surface coating with hydroxyapatite

5 Emulsions were prepared using: (A1) 78% styrene, 8% DVB and 14% Span 80 or (A2): 15% styrene, 60% 2-ethylhexyl acrylate, 10% DVB and 15% Span 8. The aqueous phase contained either; (B1): 1% potassium per sulphate or (B2): 1% potassium per sulphate 0.5% hydroxyapatite (HA) and

10 15% phosphoric acid which is used to dissolve the hydroxyapatite (HA). Emulsification is carried out at 25°C, with water phase volume at 85% and the aqueous phase is delivered through a single inlet. The processing conditions were $t_D = t_H = 600$ seconds, $R_D = 0.83$ ml/s, $R_M = 4.7\text{s}^{-1}$ and $R_E = 0.0067\text{s}^{-1}$. After the emulsification and polymerisation, samples containing

15 hydroxyapatite were soaked in 1M NaOH to precipitate hydroxyapatite. After precipitation, the samples were cleaned in water to retain pH = 7 and IPA and examined for their structure.

Table 3 - The effect of hydroxyapatite on pore and interconnect size.

Description	Oil and Aqueous Phase Compositions			
	A ₁ B ₁ (Rigid PHP)	A ₁ : B ₂ (Rigid/HA PHP)	A ₂ : B ₁ (Elastic PHP)	A ₂ : B ₂ (Elastic/HA PHP)
Pore size D (μm)	22	38	9	17
Interconnect size d (μm)	2	11	1.5	6
d/D	0.09	0.29	0.17	0.35

Samples with compositions (A₁; B₁) and (A₁; B₂) are examined under SEM
 5 and their surface elemental analysis indicated uniform distribution of
 hydroxyapatite (Ca₁₀(PO₄)₆(OH)₂) as shown in Table 4.

Oil phase containing just styrene is referred to as rigid PHP and when oil
 phase contains 2-ethylhexyl acrylate, the resulting composition is referred to
 10 as elastic PHP.

Table 4 - Elemental analysis of the fracture surface of two polyhipe polymers with compositions (A₁; B₁) and (A₁; B₂)

Component	Unmodified PHP (A ₁ :B ₁) - wt%	Hydroxyapatite coated PHP (A ₁ ; B ₂) wt%
Carbon	82	74
Oxygen	12	21.4
Calcium	-	3.2
Phosphorus	-	1.4

- 5 This in-situ hydroxyapatite coating of polyhipe is preferred since post-polymerisation soaking of the material in hydroxyapatite solution and subsequent precipitation appeared to form large but localised hydroxyapatite crystals within the polyhipe pores.

10 **Example A 4 – Effect of oil phase composition on pore and interconnect size**

- 15 An emulsion was prepared as described above, including 0.5% of oil phase initiator, 1,1'-azobis(cyclohexanecarbonitrile). Total volume and viscosity were substantially unchanged. The emulsion was processed and polymerised as described above but using conventional mixing times and rates as known in the art (Dosing rate $R_D = 0.71 \text{ ml/s}$, mixing rate, $R_M = 4.7 \text{ s}^{-1}$, mixing time $t_D = t_H = 600 \text{ s}$), and the material properties evaluated. The results are shown in Table 5.

Table 5

Sample	Pore Size, D (μm)	Hole Size, d (μm)	d/D
Control polyhipe	15.8	2.3	0.15
Polyhipe with initiator	22.7	4	0.18

Example A 5 - Effect of oil phase soluble initiator on hydroxyapatite modified polyhipe.

An emulsion was prepared using 0.5 wt% hydroxyapatite solution (pH = 2.5 adjusted with phosphoric acid) without any aqueous phase initiator. The oil phase contained 77.5% styrene, 8% DVB, 14% Span 80 and 0.5% azobisisobutyronitrile (AIBN) as oil phase initiator.

All other experimental conditions were the same as Example 3. In Table 6 the hydroxylapatite modified polyhipe is compared with the corresponding material in Example 3 (i.e., rigid/HA PHP).

Table 6

Sample	Pore Size D (μm)	Interconnect Size d (μm)	d/D Ratio
Rigid/HA	38	11	0.29
Rigid/HA AIBN initiated	30	12	0.40

The above table indicates that the presence of oil phase initiator does not substantially affect the pore and interconnect sizes. However, the elemental analysis of the polyhipe surfaces using EDAX (energy dispersive analysis

with X-rays) show that the presence of HA is more pronounced if oil phase initiator is used. These results are shown in Table 7.

Table 7

5

Component	Overall analysis on HA modified polyhipe with AIBN	Overall analysis on HA modified polyhipe
Carbon	63.09	74
Oxygen	22.49	21
Phosphorus	2.98	2
Calcium	9.70	3

The concentration of calcium ions is greater than for the control which did not contain AIBN

10 Example A6 – Effect of silica particles in aqueous and oil phases

Hydrophilic silica (Aerosil 380, particle size 7nm) was incorporated in the aqueous phase while hydrophobic silica (Aerosil R812, particle size 20nm) was incorporated in the oil phase. These silica samples were obtained from Degussa, Germany. In each case silica loading was 0.5, 1.0, 2.0 or 5.0 % by weight. Phase volume of the aqueous phase was 90% and the processing conditions were: Dosing rate $R_D = 0.375$ ml/s; mixing rate $R_M = 4.7$ s⁻¹, mixing time $t_D = 600$ s, $t_H = 1200$ s. The oil phase initially contained 15% styrene, 62% 2-ethylhexyl acrylate, 8% DVB, and 15% Span 80.

20 Hydrophobic silica was added into this mixture at various levels. The aqueous phase contained 1% potassium persulphate. When hydrophilic silica was used in the aqueous phase, there was no silica in the oil phase and the original oil phase composition was used as above. The results are shown in Tables 8 and 9.

Table 8. Effect of silica on the pore and interconnect size of polyhipe polymer when silica is present in the oil phase only.

Silica in oil phase (wt%)	0	0.5	1.0	2.0	5.0
Pore size, D (μm)	10.7	14.3	21.9	37.2	60.7
Interconnect size, d (μm)	4.0	3.2	2.9	3.2	3.43
d/D	0.37	0.22	0.13	0.09	0.06

Table 9. Effect of silica on the pore and interconnect size of polygipe polymer when silica is present in the aqueous phase only.

Silica in aqueous phase (wt%)	0	0.5	1.0	2.0	5.0
Pore size, D (μm)	10.7	17.6	28.9	36.2	Coalescence pore formation
Interconnect size, d (μm)	4.0	3.6	23.4	1.9	
d/D	0.37	0.21	0.12	0.05	

Example B 7 – Multiple cell growth in polyhipe

Chondrocytes (obtained from bovine metacarpalangeal joint) human osteoblasts cells (HOB) and 3T3 fibroblast rat cells were obtained. Two samples of polyhipe, without and with HA were cut into discs, washed and thoroughly sterilised. Thermanox discs were used as control. The discs were soaked in the respective cell culture medium for 24 hours and placed in well plates. 3T3, HOB and chondrocytes cells were each seeded onto the discs at a densities of 100,000, 100,000 and 500,000 cells per ml respectively. The plates were then incubated at 37C in 5% CO₂ for up to three weeks. Samples were examined after 1,7,14 and 21 days by SEM and for histology.

Results

The 3T3 which grew on the rigid polyhipe can be seen to have multiplied rapidly, by day 1 a confluent layer was present on both types of polyhipe.

5 This layer remained throughout the duration of the experiment. From histology it could be seen that there was no penetration, in some samples by day 7 and in all samples by day 14 the cell layer had begun to leave the surface of the polyhipe.

10 The HOB cells showed well spread morphology on day 1 under SEM, the cells were bridging the interconnects on the surface of the polyhipe, opposed to growing into the interconnects. By day 7 all samples had a confluent layer on the surface. From histology was noted that there was slight penetration, only to the depth of one pore in the samples containing HA. The layer
15 remained securely attached to the surface throughout the duration of the experiment.

The chondrocytes at time points of 1 and 7 days adhered to the polyhipe discs and showed rounded morphology on both types of polyhipe, compared
20 to the Thermanox^R control where the cells had spread out and become fibroblastic in appearance. By day 14 there was signs of flatter cells and by day 21 there were confluent layers on some samples. The cells on the HA polyhipe could be observed to flatten at early time points compared to the unmodified polyhipe. From histology it was noted that from day 5 onwards
25 there was penetration in both polyhipes and cells were present within the 3D structure, however the HA polyhipe had greater penetration and visually seemed to contain more cells. The cells which had penetrated the unmodified

polyhipec were fewer in number however were retaining their rounded morphology.

From the results of Safinin O staining it can be shown that healthy chondrocytes were present in both the centre and the periphery of the polymer matrix. There were viable cells secreting GAG both in the polymer and on the periphery.

The rate of chondrocyte growth studies (as obtained from DNA assays) indicate that the hydroxylapatite modified polyhipes yield significantly faster growth compared with unmodified polyhipes as shown in Table 10.

Table 10

Type of Assay	Cell Lysis (mg DNA/ml)	
Time (day)	PHP	HA-PHP
1	3.1	3.2
5	3.6	4.4
10	3.6	9.5

Example B8 – Effect of Pore Size on Cell Growth

Six types of styrene / 2-ethylhexyl acrylate Polyhipec Polymers (containing hydroxapatite coating) were produced with average pore size of 8, 17, 24, 31, 34 and 89 μm . The interconnect size in all cases were similar. These samples were then seeded with chondrocytes and the effect of pore size was noted. The results are shown in Table 11.

Table 11. Effect of PHP pore size on the cell penetration and collagen II production after 21 days of cell culture.

Pore size, D (μm)	8	17	24	31	45	89
Interconnect size, d (μm)	3	5	6	6	7	7
Depth of Penetration (μm)	17	508	570	367	67	45
Relative Collagen II Production	23	48	52	42	12	9

- 5 The production of Type II collagen is an indication of the correct physiological cell function. This example indicates that, for chondrocytes which have cell size of approximately 10 μm , the optimum pore size is approximately 25 μm . If the pore size is too low, the cells can not penetrate the polymer and if the pore size is too large, the cell morphology changes
- 10 from rounded to a flat and fibroblastic appearance. These fibroblastic cells proliferate rapidly and form a layer on the surface rather than penetrating the polymer.

Another measure of extra-cellular matrix production is the yield of

15 glycosaminoglycan (GAG). The concentration of GAG was determined by a colorimetric assay. The effect of pore size in the production of GAG was evaluated using the cell culture studies summarised in Table 10. The GAG production studies were also compared with the Tissue Culture Plastic (TCP). The results are shown in Table 12.

Table 12. Effect of pore size on the production of GAG after 21 days of cell culture. Results are also compared with the cell culture studies using tissue culture plastic (TCP).

Pore size, D (μm)	8	17	24	31	45	89	TCP
Interconnect size, d (μm)	3	5	6	6	7	7	0
Relative GAG Production	192	720	784	528	384	400	352

5

Example B 9 – Macrophage Activity on Sulphonated Polyhipe Polymer

The biocompatibility of the materials can be tested by exposing them to
 10 macrophages. The response of macrophages to a material is assessed by a)
 Morphological changes to macrophages, b) Hydrogen peroxide production
 by macrophages and c) Beta glucuronidase production by macrophages.
 Macrophages are spherical *in vitro* and any deviation from this response is a
 negative response. Excessive production of hydrogen peroxide and beta
 15 glucuronidase are also negative responses.

A styrene / 2-ethylhexyl acrylate Polyhipe Polymer is produced. Some of
 this material is sulphonated (degree of sulphonation is 12%) and
 subsequently neutralised using sodium hydroxide. Macrophages are seeded
 20 on these polymers and their morphological and hydrogen peroxide and beta
 glucuronidase production capacities are assessed after six hours. The results
 are shown in Table 13.

Table 13. Effect of substrate on macrophage activity

Substrate Type	Morphology	Relative Hydrogen Peroxide Production	Relative Beta Glucuronidase production
PHP	Round	370	47
Sulphonated PHP	Flat	975	684
TCP*	Round	295	93

- TCP = Tissue Culture Plastic

5

Table 13 indicates that sulphonated PHP yields very high level of negative activity and therefore such polymers may be used as negative standard in which the activity is reduced as the degree of sulphonation is decreased.

10 Example C - Fabrication of an Organ Support System

Figures 1a - 2b illustrate micro- and macroscale capillary and channel configurations in polyhipe modules obtained by the method of Example C.

- 15 In a cell support system, we can assume that only Type-1 pores (basic pores) are present with probably a set of micro-capillaries which provide an effective cell seeding and subsequently act as nutrient channels. In this case, these micro-capillary channels need not to be closely packed and they can have large diameters approaching 500 micron. Such cell support systems can
- 20 have the shell-and-tube type of arrangement and can be constructed to have a cylindrical or square or rectangular cross-section. This arrangement is similar to the hollow fibre bundles except that the shell side constitutes the bulk PHP and micro-capillary/bulk PHP interface is very thin (0.5-5micron).

In an organ support system there is more than one type of micro-capillary is present. In a cubic or polyhedral organ support, cartridge, the opposing faces can be connected to each other with micro-capillaries. For the simplest case, a cubic or cuboidal organ support cartridge with three types of micro-channels are illustrated in Figures 1a, b. Figure 1a, 3D representation of the micro-capillaries are shown (only 3 types of capillaries are illustrated) while in Figure 1b, cross-section is illustrated showing all the capillaries and PHP in the bulk. Here, the channels are denoted as A-channels, B-channels, and C-channels.

The following procedure is used to obtain a 3D-network of fibres in a mould which is subsequently filled with the emulsion of the PHP and polymerised afterwards. However this procedure is not unique but it is intended to illustrate the fabrication technique of the in vitro organ support systems.

1. Strands of desired diameter metallic rods or fibres are sandwiched between two sheets of polymer using either a hot press or two roll-mills (calendars). The metallic fibres or rods are parallel to each other and they will eventually create the A-channels. These sandwiched panels are cut to a desired size and optionally stamped with a die to form holes which will eventually house the metal rods leading to the formation of the B-channels. Several of these sandwiched panels are produced in this way.
2. The above procedure is repeated using desired diameter metallic rods or fibres in order to obtain the C-channels. If the diameters of the A-

and C-channels were to be the same, there is no need to have separate fibres/rod sandwiched panels.

3. These panels are laid in a mould on top of each other with two types of thin metallic spacers between them. The first type of spacer secures the fibers in place while the second type can be removed when the fiber assembly is compressed in the direction of the B-channels. The orientation of the fibres which will form the A- and C-channels are perpendicular to each other. The mould already contains metallic rods which will form the B-channels. The thickness of the thin metallic spacers determines the separation of the A- and C-channels.
4. These panels are securely tightened and placed in an oven which allows the tightening of the fibre/rod layers as the temperature of the oven is raised gradually above the melting point of the polymer used in the sandwiched panels. The temperature is further raised in order to burn off the polymer completely leaving a 3D network of fibres/rods.
5. These fibres/rods may require surface coating with polymer in order to create a desired capillary/bulk interface. For example, these template fibres/rods can be coated with hydrophobic polymers to obtain closed pore structure at the interface when styrene based PHP polymers are produced. If the template fibres/rods are coated with water soluble polymers such as polyethylene oxide, Type-2 pores may be obtained at the interface. When metals are not coated, open pore structure is obtained at the interface.

6. The mould is filled with the emulsion slowly under vibration in order to obtain complete filling. The mould is further compressed in the direction of the B-channels by removing the type-2 metallic spacers, thus allowing the escape of the excess emulsion. Emulsion viscosity should be as low as possible and the emulsion should be re-circulated during filling.

7. After mould filling and polymerisation, the template rods are removed and the fibres are dissolved in acid thus exposing the microcapillary channels.

After the fabrication of the organ support cartridge, it is washed in water and neutralised and subsequently washed with ethanol to remove any residual monomer and re-washed with water and dried before being sterilised ready for use as an *in vitro* organ support.

In the cuboidal modules of Figures 1a - 2b, there are three types of channels A, B, C. These are shown in Figures 2a and 2b, connecting the faces of a cube, indicated in Figure 2a as A-A', B-B' and C-C', respectively. A-channels may be carrying blood (or plasma, or nutrients) B-channels may be used to transport the cell products (cell expression, metabolic products) while the C-channels can be used to grow anisotropic cells such as nerve or muscle cells. Using a cuboidal module with eight faces and hence four types of micro-channels, one can also grow muscle cells together with nerve cells, still providing for blood and cell product collection. The space between the micro-channels will be occupied with a specific organ cell or connective tissue or tissue or stimulant.

Each channel can be made of different diameter and different packing density and they can be independently surface modified (for example made electrically conductive, or have different interface porosity or pore size, or different chemical structure). Instead of growing nerve cells, it might be
5 sufficient to have for example, carbon fibres present in these channels to provide electrical conductivity.

CLAIMS

1. Micro cellular polyhipe polymer scaffold suitable for growth of living matter for biomedical applications, obtainable by polymerising a high internal phase emulsion, comprising a homogeneous cross linked open cellular material defined by a bulk polymer matrix having a surface and an interface with an internal phase, and having porosity greater than 75% comprising emulsion derived pores of diameter in the range of 0.1 to 10,000 micron and emulsion derived pore interconnects of diameter in the range of up to 100 micron, wherein the scaffold comprises a plurality of discrete and/or interpenetrating zones:

at the polymer surface;

within its bulk matrix;

at the interface between polymer and internal phase; and/or

between adjacent but distinct pores and/or interconnects,

characterised by form and dimension of pore and interconnect type within each zone, and location of zones wherein adjacent or interpenetrating zones are distinguished by boundaries which may be between or contained within adjacent pores and/or interconnects in respective zones, whereby zones are suitable for regulating positioning and/or morphology of living matter.

2. Microcellular polyhipe polymer scaffold according to Claim 1 suitable for growth of living matter selected from cells, micro-organisms such as bacteria and virus and mixtures thereof.

3. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 2 comprising micro channels formed of pores with interconnects suitable for providing communication and penetration of living matter for anisotropic

(directional) growth thereof.

4. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 3 wherein the walls of the micro-channels are (bio)degradable suitable for fusion of living matter in the (bio)degraded scaffold.
5. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 4 comprising in individual zones, pore and interconnect sizes in different ranges, suitable for co-culturing two or more types of living matter.
6. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 5 wherein ratio of interconnect to pore diameter is in the range $0 < d/D < 0.5$, preferably in the range $0.1 < d/D < 0.5$ when the pore diameter is approximately less than 200 micron
7. Microcellular polyhipe polymer scaffold as claimed in Claim 6 comprising extensive networks of elongate micro capillaries obtainable by moulding about fibrous inserts of diameter as low as 10 micron, throughout the scaffold or zones thereof, separated by the microcellular polymer wherein microcapillaries are suitable for blood or nutrient supply channels, expression channels for living matter and seeding of living matter.
8. Microcellular polyhipe polymer scaffold as claimed in any of Claims 6 and 7 wherein the interface between a microcapillary wall and the bulk polymer provides a thin surface layer of the order of 0.5-5 micron, forming a zone particularly suited for directional (anisotropic) growth of living matter.

9. Microcellular polyhipe polymer scaffold as claimed in Claim 8 as dependent on Claim 2 wherein the interface has smaller pore size than the bulk polymer wherein the zone is suitable for growth of cells forming a lining, for example cells lining the blood vessels or for growing endothelial cells on the interface surface.
10. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 9 comprising a module of shell and tube type or cubic/polyhedral type with respect to direction and/or configuration of channels and/or microcapillaries.
11. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 10 comprising a surface coating, using coating materials introduced *in situ* during polymerisation or post polymerisation
12. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 11 wherein polymer is selected from proteins and cellulose, polyacrylamide, polyvinyl in rigid or flexible form, poly(lactic acid), poly(glycolic acid), polycaprolactone, poly (lactide/glycolide) and polyacrylimide.
13. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 12 wherein polymer comprises resiliently deformable or elastic material or is rendered resiliently deformable or elastic and is suitable for repeated stress and relaxation by means of oscillatory straining of the scaffold during cell growth facilitating rate of cell growth.
14. Microcellular polyhipe polymer scaffold as claimed in any of Claims 1 to 13 wherein polyhipe scaffold is electrically conductive or is rendered

electrically conductive whereby it is suitable for conducting an electric current during cell growth, facilitating distinguishing certain cell types and promoting growth and fusion of particular cell types

15. A process for the preparation of a microcellular polyhipe polymer scaffold as hereinbefore defined comprising in a first stage the formation of a high internal phase emulsion (HIPE) of an immiscible dispersed phase in a continuous phase, wherein the dispersed phase is void or contains dissolved or dispersed materials, and (co)monomers, oligomers and/or pre-polymers are present in the continuous phase, by means of introducing the dispersed phase by controlled dosing into the continuous phase with controlled mixing at controlled temperature and rate to achieve an emulsion of controlled pore size, and subsequently homogenising for controlled period under controlled deformation and polymerising under controlled temperature and pressure.

16. Process as claim in Claim 15 wherein controlled pore size of emulsions up to 0.5 μm are obtained using very high deformation rate flows, pore size up to 300 μm are obtained using rate just above the critical deformation rate at which phase inversion takes place, very large pore size up to 10,000 μm are obtained through the method of controlled pore coalescence during polymerisation, and nano-pore size up to nm are obtained through solvent extraction after polymerisation.

17. Process as claimed in Claims 15 or 16 comprising co-extrusion of polyhipe emulsions of differing pore and interconnect sizes eg concentrically or side-by-side.

18. Process as claimed in Claims 15 to 17 wherein the emulsion comprises aqueous and non-aqueous phases, preferably aqueous and oil.
19. A biologically active system comprising a polyhipe scaffold as defined in any of Claims 1 to 14 and living matter providing normal cell functioning associated with a natural biologically active system present in the human or animal body, wherein living matter is selected from microorganisms or multiple cells selected from human, animal and plant cells, preferably selected from isotropic tissue and bone cells present in cartilage, cornea, marrow and the like, anisotropic cells such as nerve, muscle, blood vessel cells, of cell type selected from fibroblasts, chondrocytes, osteoblasts, bone marrow cells, hepatocytes, cardiomyocytes, neurons, myoblasts, macrophages and microvascular endothelium cells.
20. Method for growth of multiple cells in a polyhipe scaffold as hereinbefore defined in any of Claims 1 to 14 comprising providing cells on or in the scaffold in a controlled environment and providing a suitable nutrient adapted for growth and providing conditions for growth promotion and positional control.
21. Use of a biologically active system as defined in Claim 19 as an implant or in association *in vivo* in or with the human or animal body or as a module for *in vitro* studies mimicking a part of the human or animal body or for use in a growth environment, for example for the growth of organ cells in the cell side of the module in order to stimulate organs.
22. An organ support module comprising a cubic or polyhedric module of

closely interwoven but not interconnecting channels immersed in a polyhipe scaffold as defined in any of Claims 2 to 14 suited for growth of specific organ cells in the polyhipe and/or the channels, wherein cells are optionally in contact with a specific microchannel and all cells are capable of intercell communication.

23. A method for manufacturing an organ support module as defined in Claim 22 comprising providing a polyhipe emulsion, providing a mould including inserts such as rods or fine fibres, pumping polyhipe emulsion and optional filler into the mould about the inserts and polymerising, with subsequent removal of inserts and optional filler to provide pores and interconnects, microcapillaries and nano pores in desired configuration.

24. The use of a polyhipe scaffold, a biologically active system, or organ support module as defined in any of Claims 1 to 14, 19 and 22 for the manufacture of contact lenses, dental fillings, cochlea implants, vascular supports including heart valves and cardiac pace makers and drug delivery skin patches.

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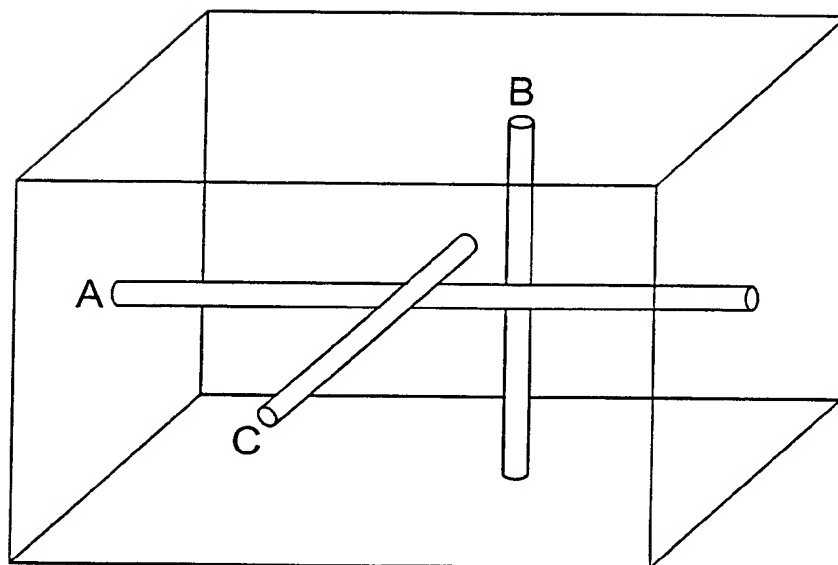
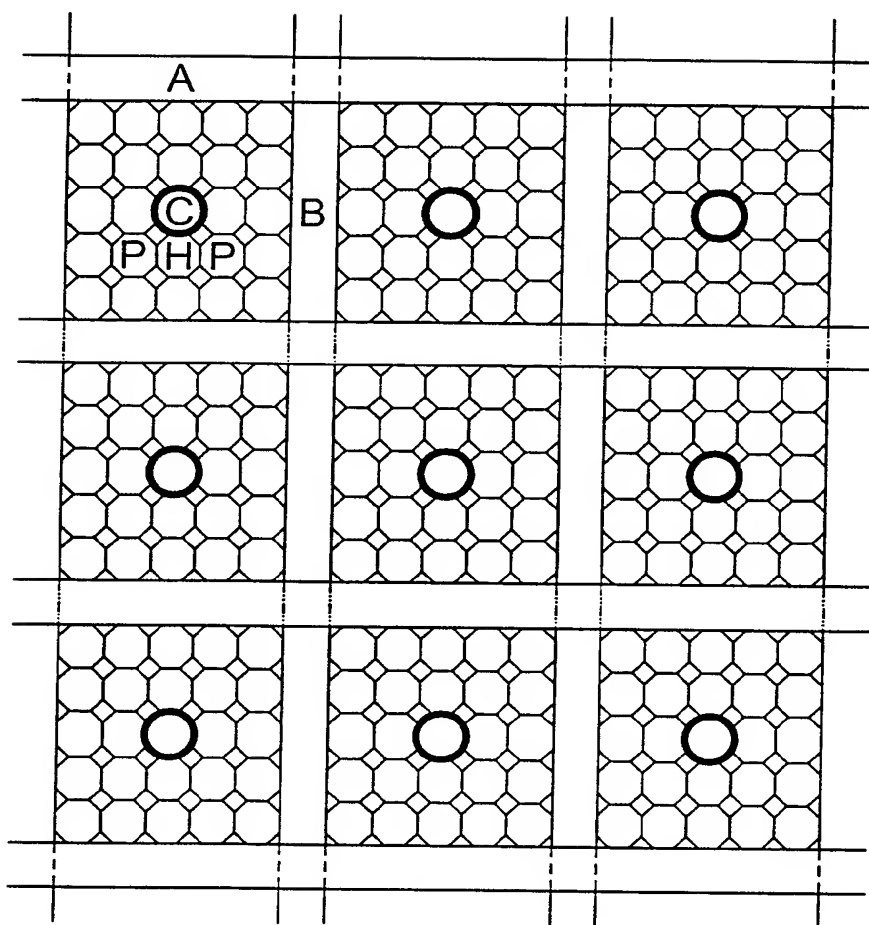
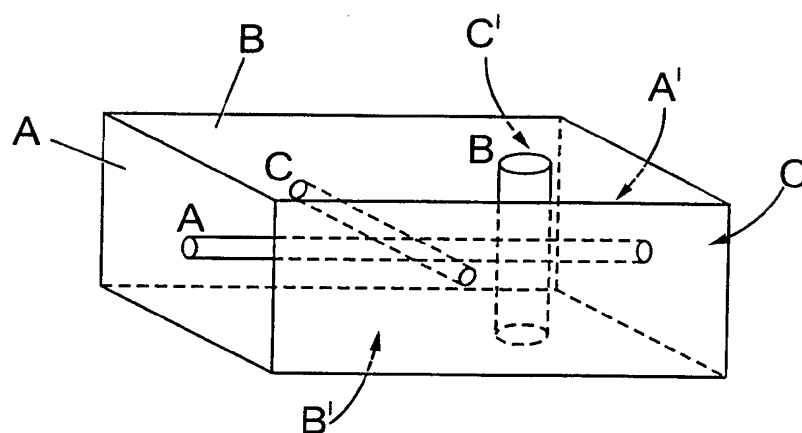
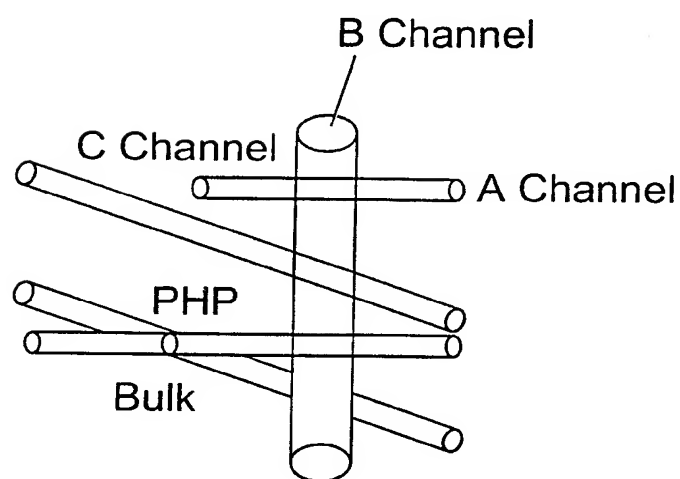


Fig. 1a

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*Fig. 1b*

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*Fig. 2a**Fig. 2b*